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TEMPERATURE MEASUREMENT AND MONITORING DEVICES

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This report has been reviewed and is approved for publication.

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The principal aim of this study was to evaluate current literature on noninvasive methods and instruments on core temperature (body's temperature in deep tissue, i.e., heart). A specific goal is a conceptual design for a device to measure core temperature through chemical defense clothing.

This report reviews temperature sensing devices, infrared thermography and emphasizes microwave thermography (theory, characteristics and relationship to measurement in the body). A conceptual design and feasibility based on potential usefulness in clinical medicine was explored. All information herein was obtained from literature review only. No laboratory research was conducted on this project. — Keywords included

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The conclusion reached is that a stand-off microwave core temperature measurement is feasible, but more research and development is required before such a device can be incorporated into a vital signs monitor.

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TEMPERATURE MEASUREMENT AND MONITORING DEVICES

INTRODUCTION

Human body core temperature, under normal environmental conditions, remains constant at 37 °C (98.6 °F within ±1 °F). Even when exposed to outside temperatures ranging 13 °C to 60 °C (55.4 °F to 140 °F), the body's core temperature will remain virtually constant at 36.7 °C to 37 °C (98.0 °F to 98.6 °F) [1]. Because the body's core temperature is so stable under normal conditions, any fluctuation from normal is a good indicator of illness or abnormality. The purpose of this report is to provide a literature review of developments in temperature measurement and in monitoring devices with emphasis on advanced methods. Major areas included in this report are: historical contributions in body temperature measurements, applications for temperature measuring devices, and description of several modern body temperature monitoring devices (techniques). Finally, state-of-the-art and potentially useful temperature devices are discussed in the report.

BACKGROUND

HEAT PRODUCTION AND HEAT LOSS

Heat is produced in the body as a by-product of metabolism and is lost continuously to the environment. Five principal factors play a major role in the rate of heat production: (1) basal metabolic rate of all cells; (2) change in the rate of metabolism by muscle activity (including the shivering reflex); (3) change in metabolism caused by the effect of thyroxine on cells; (4) change in metabolism caused by the effect of norepinephrine and sympathetic stimulation on cells; (5) change in metabolism caused by an increase of body temperature. The body loses heat through radiation, conduction, convection, and evaporation [1]. The major components in the thermoregulation of the body include an insulator system, a heat transfer system, a temperature sensing system, and a regulation system [3].

The skin, the subcutaneous tissue, and the fat in subcutaneuous tissue are the body's heat insulators. Fat is the most efficient tissue heat insulator since it conducts only about one third as much heat as the other insulators. The insulator system beneath the skin is an effective means for maintaining internal temperature, while the temperature of skin approaches the environmental temperature [2,3].

Blood vessels penetrate the subcutaneous insulator tissues and are distributed in the subpapillary portions of the skin. Immediately beneath the skin, a continuous venous plexus is supplied by blood. The flow of blood into the venous plexus can vary from 0-30% of the total cardiac output. A high blood flow conducts heat from the internal portions of the body to the skin with great efficiency; whereas reduction in blood flow decreases the efficiency of heat conduction

from the internal protions of the body. Thus, the skin acts as an effective "radiator system", and the flow of blood to the skin is an effective mechanism for transferring heat from the body core to the skin. Heat conduction to the skin is controlled by the degree of vasoconstriction of the arterioles that supply blood to the venous plexus of the skin. In turn, vasoconstriction is controlled primarily by the sympathetic nervous system in response to changes in either the internal or the environmental temperature [3,4].

Body temperature is regulated by nervous feedback mechanisms. Most of these feedback mechanisms operate through the temperature regulating center located in the hypothalamus, where warm and cold receptors detect internal (core) temperature change [4].

The preoptic area in the hypothalamus contains large numbers of heat sensitive neurons which seem to play an important part in body temperature regulation. The firing rate of these neurons increases as the temperature rises. Cold sensitive neurons exist in other parts of the hypothalamus, in the septum, and in the reticular substance of the mid-brain. These neurons increase their firing rate as the temperature decreases, but they play only a minor role in body temperature regulation since there are so few of them compared to the heat sensitive neurons. Experiments have shown that as the preoptic area is heated, profound sweating results and the blood vessels dilate to radiate heat from the body. This reaction lowers the internal body temperature to a normal level. In conjunction with this mechanism, body heat production is inhibited; i.e., cell metabolism decreases [4].

The heat sensitive neurons in the preoptic area can also detect cold temperature to some degree by decreasing their firing rate as the temperature decreases. However; by the time the internal body temperature has fallen a few tenths of a degree below normal, the firing rate is reduced to the extent that the heat sensitive neurons become completely inactive. Therefore, cold detection must rely on the cold receptors of the skin, spinal cord, abdomen, and other internal structures of the body. The skin has both warm receptors and cold receptors, but there are about 10 times as many cold receptors as warm receptors. Therefore, peripheral temperature detection is concerned primarily with detection of cold temperature. When the skin is cooled, the cold receptors increase firing rate and produce a reflex to raise the temperature of the body by several mechanisms: (1) providing a strong stimulus to cause shivering, (2) inhibiting sweating, (3) promoting vasoconstriction of blood vessels in the skin to diminish the transfer of heat from the body core to the skin [2,5].

Thus it is apparent that the preoptic area of the hypothalamus is mainly concerned with preventing hyperthermia, overheating of the body; and the cold receptors, primarily in the skin, are mainly concerned with preventing hypothermia, subnormal body temperature.

Signals from the cold receptors and from the preoptic area in the hypothalamus stimulate an area located bilaterally in the posterior

hypothalamus approximately at the level of the mammary bodies. It is here where these two signals combine to provide either a heat producing or a heat losing reaction [2].

Shivering begins at a hypothalamic temperature of approximately 37.1 °C (98.7 °F) when the skin temperature is at 20 °C (68 °F). When the skin becomes cold it drives the hypothalamic thermostat to the shivering threshold even when the hypothalamic temperature is still quite warm. Thus, this mechanism of cold skin temperature increasing heat production actually "anticipates" a possible fall in the internal body temperature and tries to prevent its occurrence. The primary motor center for shivering is located in the dorsomedial portion of the posterior hypothalamus near the wall of the third ventricle. The motor center for shivering is normally inhibited by heat signals from the preoptic area, and is driven by cold signals from the skin and the spinal cord. When this center becomes activated it transmits impulses down to the anterior motorneurons. These nonrhythmic impulses do not cause actual muscle shaking. Instead, they increase the overall tone of the skeletal muscles throughout the body. When the muscle tone rises above a certain level, shivering begins, which is probably the result of feedback oscillations of the muscle spindle stretch reflex mechanism [4].

CLINICAL APPLICATIONS OF BODY TEMPERATURE MEASUREMENT

The human body has an extremely effective thermoregulation system that regulates the body core temperature to ±0.55 °C (1 °F) from a preset normal. Many physiological processes enter into this mechanism -- perspiration, vasodilation, vasoconstriction, shivering, and piloerection among others. Unfortunately, the body's thermoregulatory system for controlling these processes is still not fully understood. What is understood is that deviation from normal body temperature is an excellent indicator of a body in abnormal condition.

The body responds to bacterial infections and diseases with an increase in core temperature (fever). By increasing the body temperature, bacterial growth decreases, thereby helping the immune system destroy the invaders. Damage to the thermoregulatory center in the hypothalamus may also cause fluctuations of body temperature.

Severe effects due to environmental conditions can drastically change body temperature. Exposure to extreme heat may cause a condition known as heat stroke. Body temperature in this case will rise to between 41 °C and 42.2 °C (106 °F and 108 °F). At this temperature, damage to the parenchyma of cells begins. If exposure continues, great damage to neuronal cells and those of the liver and kidney will eventually cause death. Therefore, it is exceedingly important to continually monitor a patient with an elevated body temperature.

Other applications for temperature monitoring is in the observation of breast cancer, peripheral vascular disorder, trauma and wound

healing, orthopedics, arthritis, dermatology, obstetrics and gynecology, drug therapy, and ophthalmology.

TEMPERATURE SENSING DEVICES

Hippocrates is believed to be the first person to associate body temperature as sensed by his hands with a person's well being. Unfortunately, he concluded incorrectly that fever was a type of disease rather than an effect. Before Hippocrates, it was believed that fevers were caused by angered gods. The use of temperature measurements as a diagnostic tool was delayed for several more centuries by the lack of understanding the body's thermoregulatory mechanism.

Galileo is credited with inventing a thermoscope, the first temperature sensing device. Unfortunately, the thermoscope lacked a reference scale and was not a useful temperature measuring device. Sanctorius of Padua was the first to fabricate an air thermometer to measure the body temperature of a human with fever. In 1630, Jean Rey, a French physician, invented an open-system liquid thermometer which depended on barometric pressure and, therefore, could not easily make precise measurements.

In 1693, Halley and Boyle both suggested the use of mercury in glass tubes because the mercury remained in liquid form over a large temperature range without expanding or freezing. The reasons for using mercury in the thermometer included low freezing and high boiling points (-40 °C and +357 °C [-40 °F and +674.6 °FF respectively), rapid heat conduction, large uniform expansion with rise in temperature, and good visibility. As the mercury in the reservoir is heated, it expands up a capillary tube to a level for a determined temperature. The mercury thermometer has several limitations and disadvantages. If the capillary diameter is too large, the mercury tends to return toward the reservoir and indicate a lower than actual temperature. If the capillary diameter is too small, the mercury tends to rise rapidly up the tube instead of increasing steadily. The response time for a mercury thermometer is about 90 seconds, because of the slow heat conduction in the body cavities. Mercury thermometers have an inherent error of up to 0.5 °C (0.9 °F) at normal body temperature; the error increases at higher temperatures. Another disadvantage is that the glass thermometer is very fragile. Patient, clamping down on glass thermometers by their teeth could be injured by broken glass and could injest toxic mercury. The size of thermometers is limited and temperature may be taken at just a few body locations.

Mercury thermometers cover a wide temperature range, but until the early 1700's, there was no standard scale. Gabriel Fahrenheit established a standard scale based on the freezing and boiling points of water. In addition, Fahrenheit developed a procedure to distill mercury to prevent it from adhering to glass surfaces. A Swedish astronomer, Anders Celsius, offered another solution with a scale based

on the same properties of water, but he separated the temperature range into one hundred divisions. Linnaeus modified the scale developed by Celsius resulting in what is now known as the Centigrade scale. It was not until 1776, that physicians routinely used the thermometer in patient examinations; and not until the middle 19th century, that Wunderlich finally demonstrated that the fever Hippocrates thought was a disease was, in fact, just a symptom of a much more complex illness.

In the 19th and 20th centuries, new methods of temperature measurements were developed. In 1800, Sir William Herschel discovered infrared radiation, but it was not until 1840, that his son, Sir John Herschel, produced the first crude thermograms. In 1825, Seebeck developed the thermocouple. In 1929, Professor Marianus Czerny developed what was the beginning of thermography, a thermal imaging system [6]. Czerny evaporagraphy made heat patterns visible through a variable rate of evaporation of alcohol on specially treated paper. In 1957, it was shown that breast tumors could be detected by a rise in temperature with respect to the surrounding tissue [7].

Temperature measured with a resistance coil is based on the change of wire resistance with changes in temperature. Platinum is preferred because of its anticorrosion properties and its large coefficient of resistance (0.004/°C). Temperature is calculated from the equation formula:

$$R_t \text{ is } R_0(1 + at) \qquad (1)$$

where:

t is measured temperature
a is coefficient of resistance of the particular metal (constant)
R_0 is resistance at 0 °C (constant)
R_t is resistance at t °C (measured)

Temperature resistance coils operate over a wide temperature range, from -200 °C (-328 °F) to 600 °C (1112 °F), and are accurate to ±0.001 °C (.0018 °F). Early temperature resistance devices were bulky, but through metallurgical advances, the size is decreasing.

The thermistor is another temperature measuring device based on the electrical property of changing resistance with changing temperature. Thermistors are small and extremely sensitive. Usually their temperature range is kept well within -80 °C (-112 °F) to +150 °C (+302 °F) in order to retain sensitivity. Thermistors exposed to temperatures outside their normal operating range must be recalibrated, but thermistors operated within their design range will have a long life and remain accurate for many years.

The thermocouple, based on the Seebeck effect, is another electrical temperature sensing device in which two dissimilar metals at different temperatures create a current proportional to the temperature difference between the junctions of the dissimilar metals. Thermocouples are small, versatile, and fast acting. In temperature measurement, one of the junctions must be maintained at a constant reference temperature; the accuracy of most thermocouples is within ±0.1 °C (.18 °F).

Liquid crystals can detect temperature changes as small as 0.3 °C (.54 °F). A strip of chloresteric liquid crystals is placed on the body surface to monitor body temperature changes. Relative temperature is indicated by a color change. This method of measuring temperature is used primary to monitor "temperature-trend" instead of actually measuring body temperature. Commercially available strips can measure temperatures within the range of 34.5 °C to 40.0 °C (94.1 °F to 104 °F).

Another method of measuring temperature is with the use of magnetic fields. The magnetic flux density of certain alloys changes inversely with temperature. The alloy is bonded to a silver plate which then is placed in contact with the skin. Magnetic temperature measurement has been used clinically, but its use is not as widespread as standard thermometers.

All of the early methods of temperature measurement require contact between the sensing element and the body. A noncontact thermometer called a radiometer senses radiant energy from the body to make relative surface temperature measurements. This method of temperature measurement is termed thermography, a graphic representation of infrared radiation from the body. The basis for thermography is property of all objects to emit and absorb energy. The human body loses over 60% of its heat through radiation. Heat loss also occurs through convection, conduction, and evaporation -- heat loss mechanisms which are important since radiation requires a temperature gradient, i.e., radiation from the body occurs if the body is warmer than the environment [8].

Thermograph systems have two main units; a camera unit and a recording/display unit. The camera unit includes a scanning element with a rotating multifaceted mirror, an optical system, and an infrared detector. The infrared detector transforms radiant energy into an electrical signal. Commercial detectors use either indium antimonide (InSb) which can sense infrared radiation up to 5 μm, or mercury cadmium telluride (HgCdTe) which can sense infrared radiation up to 10 μm [6], and liquid nitrogen for cooling.

The critical consideration in the application of any infrared thermograph system includes the field of view, emissivity of the surface being measured, atmospheric effects, and spectrum response or temperature range to be measured.

The field of view is the angle of view which is determined by the thermographic system's optics. Since the infrared unit integrates and averages the temperature of all surfaces within the field of view, the measurement will not be accurate if the target surface does not completely fill the field of view, especially if the background temperature differs from the target temperature [9].

Emissivity is defined as the ratio of the energy radiated by an object at a given temperature to the energy emitted by a perfect

radiator (blackbody 1.0). All values of emissivity fall between 0.0 and 1.0. The total radiation energy of a body is the sum of the energy emitted as a result of the temperature of the body, energy from other sources reflected by the body, and energy from other sources transmitted through the body. The higher the emissivity of a body the easier it is to obtain an accurate temperature measurement; however, reflectivity will affect accuracy. Reflectivity of the body is a more important consideration than transmissivity [4].

Thermographic measurements with an infrared sensor are made at a distance from the object (subject) and may be obtained while the subject is moving. Because thermography is a noncontact passive measurement, there is no danger from harmful ionizing radiation, as there is with X-rays.

Thermography has been used primarily for the detection of breast tumors, although it can be used on any part of the body. Thermography can be used in tumor detection because tumorous tissues usually have a higher metabolic rate and a higher temperature than surrounding healthy tissue. Unfortunately, thermograms cannot discriminate between benign and malignant tumors, and some small tumors do not produce enough heat to be noticed by the infrared thermograph. Although these limitations reduce its usefulness in breast cancer detection, thermography is a valuable screening tool [7]. The use of infrared thermography is limited to the determination of skin surface-temperature and does not necessarily represent the body's core temperature.

MICROWAVE AND THERMOGRAPHY

Studies of electromagnetic radiations from biological specimens and more recently from live human tissues have grown rapidly in the last two decades [10,11]. The most recent flurry of biological research is concerned with microwave energy. Three basic types of biological research involving microwave energy are absorption, scattering, and generation [10].

Most research efforts are directed at the absorption processes of various tissues when microwave energy is transmitted into the tissue and converted into heat [12]. These studies have shown that microwave thermography is possible. Since, absorption studies usually require powerful microwave generators, the principal aim of these studies is to establish safety standards for therapeutic and diagnostic treatments of patients with medical microwave instruments [13,14].

Other studies involve microwave scattering from biological tissue [11,15]. Griffin [11] states that about half of the microwave power incident on the surface of a subject is scattered. These scattered waves provide information on surface movement. The microwave reflection method of measuring surface movement of a subject has the advantage of not requiring any attachment or contact with the subject. The study of scattered or reflected waves requires the least elaborate

microwave measuring equipment of the three types of microwave studies. The scatter study is less expensive and safer than other microwave studies because the equipment uses a simpler generator and receiver system and because the microwave power necessary to study reflected waves of the human body is well below accepted safe emmission standards.

Griffin used a microwave interferometer system based on the three-port circulator to detect small movements through the difference in phase between an input (incidence) reference wave and a reflected wave from the target area. A square-law detector was used in the interferometer to produce a DC output current given by the law of the cosine equation:

$$I = K(e_R^2 + e_S^2 + 2e_R e_S \cos\phi)/2 \qquad (2)$$

where: K is a proportionality constant,
e_R is the magnitude of the reference signal,
e_S is the magnitude of the reflected signal,
and ϕ is the phase difference between the reference and phase signals.

The e_S^2 term can be neglected for longitudinal movements that are small compared to the distance between the test subject and the receiving antenna. The cosine term will cause an interference pattern in the detector output that decreases in amplitude with increasing separation of target and antenna. Small movements can be detected by placing the target between a maximum and an adjacent minimum in the interference pattern. The cosinusoidal interference pattern can be approximated by a straight line in this area. Movements of the target would be interpreted as a change in the detector output multiplied by a constant of proportionality. This constant of proportionality depends on the range and the slope of the line approximating the interference pattern at the operating point $\phi_o = \pi/2$ or $3\pi/2$.

The straight line approximation places an upper limit to accurate motion detection. At one-tenth of the wavelength of the microwave, the straight line approximation is no longer valid. A nonlinear scale would allow for measurements beyond the cutoff. Two devices that could be used to study motion by the microwave method were discussed by Griffin [11]. One system which is easy to operate is the Magic Tee device. Another system, the three port circulator device, was 6 dB more accurate for the same power source, but it was more difficult to operate. Griffin claims useful measurements of the chest wall motion indicating the breath rate. Griffin also claims that thermographs of movement of the foot while the knee was being supported by the other knee showed the heart rate. The suggestion was offered that the graph of the previous case could hold information of the reflex action of the subject, as well as the weight of the limb [11].

In other studies, Pedersen et al. [16,17] investigated the use of microwave radiation for pulmonary diagnosis. Their results indicated

that the amplitude and phase of reflected microwave radiation is correlated with respiratory function. In addition, the ratio of lung tissue plus fluid to total lung volume is changed by pulmonary edema and emphysema.

These results stimulated studies by Lin [15] and Iskander et al. [18-22]. Iskander et al. conducted in vivo experiments on dogs and developed a numerical procedure to simulate and verify the sensitivity of the microwave method to quantify small changes in lung water content. Iskander et al. [21] reported a close correlation between changes in the phase of microwave transmission coefficient and the pulmonary arterial pressure. Experiments on isolated perfused lungs confirmed the sensitivity of the microwave method and revealed a correlation between the phase of the microwave transmission coefficient and the weight of the isolated lung [19]. Iskander et al. [18] reached the conclusion that "nonuniformities in water distributions in the aperture of the microwave transmitter, and location on the receiver significantly affect the microwave measurement."

In the studies of microwave generation by living biological systems, very little work has been reported. These studies usually require sophisticated receiving units because of the low power of the generated waves [11]. Myers et al. [13] report using a microwave radiometer system to detect temperature gradients in live tissue. This technique, known as microwave thermography, detects the intensity of the thermal radiation emitted by the body at microwave frequencies. The microwave intensity measurements correspond to temperatures. This information is plotted in two dimensions and used to locate abnormal heating or cooling spots of the body.

Detection of thermal patterns by microwave thermography depends upon two factors: thermal radiation by the body, and the partial transparency of the tissue to microwaves. It is well established that all objects including the body emit electromagnetic radiations resulting from collisions of accelerated electrical charges caused by internal thermal motion [12] and absorb electromagnetic radiations from their surroundings [13]. The intensity of emitted radiation is approximated by the Rayleigh-Jeans equation

$$I(f,T) = e(f)*(2KTf^2)/C^2 \qquad (3)$$

where: K is Boltzmann's constant,
C is the speed of light,
T is the temperature of the object,
f is the microwave frequency,
and e(f) is dimensionless emissivity ≤ 1.

The emissivity depends upon the dielectric properties of the emitting and receiving media [12].

At normal human body temperature, the graph of specific intensity of black-body thermal radiation versus frequency has a maximum value of

about 3.0×10^{13} Hz or a wavelength (λ) of about 10 μm. This frequency is in the infrared range. At the microwave frequency of 3.0×10^{9} Hz (3.0 GHz or λ=10 cm) the electromagnetic radiation intensity emitted by the body is decreased by about 10^{8} from its peak value. Thus, expensive, advanced, low-noise microwave radiometers are essential; however, the benefit of the measurement system is that at microwave frequencies the intensity is directly proportional to the temperature of the tissue. This direct relationship is what makes the microwave thermography technique viable.

The second factor, which deals with the transparency of tissue to microwaves, is also of great importance. The effective transparency of the body to microwaves depends upon frequency and upon the properties of the tissue. It is the transparency of tissue to microwave radiation which enables temperature information to be obtained at the skin surface from internal (core) body temperature [24,26,27]. Tissue transparency is related to the dielectric properties which is largely attributed to the water content of the body tissue. Studies have shown that absorption at specific frequency depends mainly upon the water content of the tissue, and that absorption increases with frequency.

The depth at which microwave radiation can be detected depends on the emitting frequency and the tissue it is traveling through. Microwaves penetrate farthest in areas of low water content at low frequencies. However, if the frequency is too low, the penetration depth will possibly be much greater than the region under investigation, and the detected temperature variation will be reduced due to the inclusion of an increased amount of "normal" tissue.

Using these facts, Myers et al. [13] selected an operating frequency of 3.3×10^{9} Hz, giving a penetration depth of approximately 3 cm (1.2 in). In experiments, Myers et al. used a radiometer with three antennas to measure the emitted radiation. The antennas were open-ended, dielectric-filled waveguides, which were placed flush against the skin for maximum detection. Coaxial cable carried the signals from the antennas to the radiometer, and a strip-chart recorded the data. The results of the experiments showed that subsurface temperature abnormalities can be detected by microwave thermography even when surface temperature measurements show no abnormality. One experiment detected a subsurface temperature rise in the thigh of a cat caused by focused ultrasound, whereas, surface temperature measurements did not detect the temperature change. Temperature rise due to muscular contraction was detected in a human study, from which the muscle primarily responsible for the action was distinguishable. Further experiments have detected cooling as well as heating [13].

Great potential for microwave thermography is evident in the possible clinical applications. The method can be used extensively in diagnosis, such as detection of breast cancer. Cooling of the facial tissue due to cerebrovascular blockage could be detected as well as heating due to inflammation, e.g., appendicitis. However, clinical

studies and evaluation are still required in the use of microwave thermography [13].

Myers et al. [13] point out that the "resolution properties of any antenna depend upon the frequency, the antenna aperture dimensions, the source-aperture distance, and the dielectric properties of the medium being viewed." In most studies, rectangular waveguide antennas in direct contact with the skin are used. Luedeke et al. [23] indicate that microwave radiometry measurements suffer from variable emissivity mismatch between the specimen under test and the receiving antenna. Finally, although the microwave measurements are affected by several factors other than changes in lung water contents, results are encouraging. It may be possible to develop a system and method suitable for clinical application.

BASIC ELEMENTS OF MICROWAVE THERMOGRAPHY SYSTEMS

There are three basic elements of microwave thermography systems necessary to measure tissue temperature: a microwave antenna, a microwave radiometer, and a thermography output display [24]. Microwave antennas and radiometers are discussed in the report but thermography output will not be addressed.

MICROWAVE ANTENNAS

Just as radio antennas receive radiowaves to be processed and amplified, microwave thermography system antennas receive microwaves emitted from tissue. Two types are generally used: body contact antennas and remote "at-a-distance" receptors. Many factors affect the antenna's performance including its aperture size, aperture shape, and antenna composition. Perhaps the most limiting factor of antenna performance is tissue-antenna impedance mismatch. Tissue-antenna impedance mismatch is the difference in material properties of tissue and the antenna which causes signal loss, just as light waves bend and reflect when traveling from water to air. In this case, impedance is more technically defined as the measure of opposition to microwave transmission. Microwave antennas should be designed so that microwave energy loss is minimized.

MICROWAVE RADIOMETERS

The microwave radiometers measure the microwave signal's power and process it to obtain a usable temperature reading. The microwave power signal is sometimes referred to as "temperature noise". Most radiometer designs are based on Dicke's switched radiometer [25], where the source, or tissue, microwave "noise" level is compared to a reference signal of a known "temperature noise level" within the radiometer. The difference between the source temperature signal and the reference temperature signal is used to determine the temperature of the source, or tissue. In a microwave thermography system, the antenna, and radiometer interact to influence performance.

SYSTEM REQUIREMENTS

Three system parameters are important to a microwave thermography system user: temperature resolution, source spatial resolution, and tissue penetration distance. In general, each parameter has accepted standard requirements, but they may vary depending on system design and application. The power intensity of microwaves received at the antenna is a function of the source (tissue) temperature, i.e., tissue at 34 °C (93.2 °F) emits less microwave power than tissue at 35 °C (95 °F). A thermography system must be able to detect relatively small changes in microwave power, since relatively small variation in tissue temperature may indicate abnormality. It is generally accepted that microwave thermography systems should be capable of resolving temperature differences on the order of 0.1°C [24,26,27].

SOURCE SPATIAL RESOLUTION.

Microwave thermography systems must be able to detect temperature at discrete tissue areas. Depending on the application, a system should be able to resolve about 1 cm (.4 in) of tissue space.

TISSUE PENETRATION DISTANCE

The ability of microwave thermography to noninvasively measure temperature at depth is its greatest asset. In general, low frequency microwaves penetrate deeper than higher frequency microwaves, while low water content tissue (bone, fat) is more "transparent" to microwaves than high water content tissue (blood, muscle). Studies show that a microwave thermography system should operate somewhere in the range of 0.5 to 10 GHz to measure at any clinically useful depth [25].

With an understanding of the basic elements and operating requirements of microwave thermography systems, an analysis of different system designs and applications can be undertaken to determine the feasibility and desirability of developing a single, versatile system.

Microwave thermography system performance depends on the radiometer design, the antenna design, and their interaction. In the following sections, several component designs are analyzed in their respective medical applications. Analysis will offer insight into the feasibility of incorporating optimal design parameters into a single system.

MICROWAVE RADIOMETER DESIGN

OPERATING FREQUENCY: SINGLE FREQUENCY

As mentioned previously, microwave penetration distance is tissue and frequency dependent. Also, radiometer operating frequency, or the frequency at which waves are received, influences the spatial resolution of the temperature pattern. Numerous researchers have

investigated the performance of single frequency microwave thermometers used in various capacities.

Land [27] chose 3.0 GHz as the operating frequency in his microwave thermography system to achieve adequate penetration depth and spatial resolution. Choice of the operating frequency is based on the following logic: at 35-37 °C (95-98.6 °F), the loss of microwave signal to internal tissue reflection is minimum, particularly in high water content tissue, in the 3.0-3.5 GHz region [27]. Land's radiometer performed well in appendicitis detection, inflammatory joint temperature measurement, and breast tumor detection. However, the 3 GHz operating frequency is a compromise between spatial resolution (best at higher frequencies) and penetration depth. This frequency may not be ideal for a versatile microwave thermography system.

In contrast, some researchers have chosen operating frequencies tailored for temperature retrieval in specific applications. Gustov [28] investigated deep (5-7 cm [2-2.75 in]) craniocerebral temperature measurement at 1 Ghz and Thouvenot [29] used four different radiometers operating at 9 GHz, 11 GHz, 30 GHz, and 68 GHz to detect sub-cranial temperature anomalies at shallow depths. Fraser [30,31] achieved an approximate 4 cm (1.57 in) penetration depth with a 3 GHz operating frequency to monitor rheumatic disease. In breast cancer detection, Myers [32] found that a 1 GHz operating frequency had a somewhat higher true-positive (correct) cancer diagnosis, tumor detection rate than with 3 GHz. Each of the preceding cases demonstrates the necessity of having different operating frequencies for different applications.

OPERATING FREQUENCY: MULTIPLE FREQUENCY

Other researchers have used multiple operating frequencies to determine temperature profiles at discrete points. Whereas single frequency radiometers measure a weighted temperature average over a tissue thickness from the skin to a particular depth, Mizushina [33,34] developed a microwave thermography system capable of simultaneously analyzing adjacent tissue layer temperature. Tis ithermography is accomplished with an antenna which receives simultaneously microwave radiations at 1.5 GHz, 2.5 GHz, and 3.5 GHz; three separate microwave radiometers process the various frequency signals. Although, this configuration is useful, it is probably not as practical as single frequency temperature retreval, because the multiple frequency microwave system design is more complex.

SELF-BALANCING REFERENCE TEMPERATURE

As stated earlier, the source temperature is determined by comparing its microwave noise signal to the noise of a reference signal in the radiometer at a known temperature. Because of amplification variation, the greater the difference between the source temperature and the reference temperature, the greater the potential for error in the true temperature reading. In many cases, the reference temperature signal must be preset as close as possible to the source temperature to

minimize the measurement error [26]. Presetting presents a problem for many thermography systems since the source temperature (the human body) varies between subjects and within a subject.

To compensate for this source-reference temperature difference error potential, researchers have used variable reference temperatures in their radiometers. This means that when the source temperature noise signal is detected, the reference noise signal is adjusted until the difference between the two signals is zero. The temperature of the source is then known to be equal to the reference. The self-balancing reference noise used is an automatically adjusting diode. Self-balancing reference temperature signal is used in most versatile thermography systems since variability in tissue temperature is a certainty. Paglione and Mizushina [33-35] documented microwave themography systems which performed well.

NOISE-INJECTION TO MINIMIZE TISSUE-ANTENNA IMPEDANCE MISMATCH

Tissue-antenna impedance mismatch causes varying degrees of signal reflection at the skin-antenna interface. Typically, radiometers are equipped with microwave antennas composed of dielectric (non-conducting) material matching the impedance of the skin as closely as possible (the greater the impedance mismatch, the greater the error). To reduce this error, researchers have incorporated a feature which compensates for any source signal reflection [35,36]. Basically, an electronic noise signal, coupled with the antenna, is transmitted, or injected, into the tissue. The impedance mismatch will cause some of the electronic noise to be reflected back into the radiometer for processing. The reflected electronic signal is proportional to the reflected source signal, and accurate temperature measurements can be made independent of tissue impedance variation due to measurement location.

MICROWAVE ANTENNAS

MINIMIZING IMPEDANCE MISMATCHING

Typically, tissue impedance falls in the range of 50-100 ohms(Ω) [26], depending on the water content of the tissue being measured and on the moisture level of the antenna contact area. Land [26] suggests using an 80 Ω impedance antenna to compromise for the variability in tissue impedance; however, the use of this antenna will not eliminate reflection. Since the dielectric properties of the antenna cannot be adjusted, noise-injection seems to be the best alternative in minimizing impedance mismatch errors.

ANTENNA GEOMETRY

The size of the antenna aperture, independent of the radiometer operating design, affects the microwave retrieval performance. Cheever [37] tested the effect of aperture size on penetration depth in ethanol

and water. In general, he found that greater aperture sizes retrieved thermal information at a greater depth than smaller apertures. However, Prionas [38] noted that with a larger antenna aperture, spatial resolution decreases. Depending on the application, it is apparent that an optimal size is necessary.

TISSUE COOLING

Since temperature retrieval at a single operating frequency is an average over the tissue thickness, thermal conduction between a skin-contact antenna and the top tissue layer will introduce error into the entire temperature calculation. Myers [32] inserted a thin sheet of insulation between the antenna and skin to successfully prevent disturbance of the tissue's natural thermal distribution. On the other hand, Miyakawa [39] reported a prototype antenna in early development with a temperature control system that senses the skin temperature and heats or cools the antenna accordingly. The main drawback to this design is that the location of the thermistor, or temperature sensor, is 10 mm (.4 in) from the actual waveguide, which will introduce measurement error.

REMOTE SENSING

Abdul-Razzack and Butakov [40,41] have developed systems to sense microwave power emissions from the body without body contact with the antenna. With these systems, microwave signal loss is more pronounced because of additional microwave refraction and absorption in the air space between the source and the antenna. Consequently, these systems sacrifice adequate penetration distance and spatial resolution. However, remote microwave vital sign detection, including thermography, will definitely be an area of future interest and development.

CONCLUSION

It is feasible to design a microwave thermography system for use in various medical temperature sensing applications. Land and Paglione reported successful performance of "versatile" thermography systems [27,35], but versatility is limited in these systems since compromises in design were made. Actually, any microwave thermography system can be considered versatile. Its performance under a variety of conditions and its ability to be used in many circumstances determine its degree of versatility. For this reason, it can be concluded that some of the following design parameters should be incorporated into a highly versatile system:

(1) An adjustable operating frequency to retrieve microwave signals at various depths in different tissues and to achieve varying spatial resolution for different applications.
(2) A self-balancing reference temperature in the Dicke-type radiometer to minimize measurement errors.
(3) A noise-injection feature to help minimize tissue-antenna impedance mismatch.

(4) An insulated skin contact antenna to minimize thermal conduction between antenna and skin.

It is conceptually feasible to design and fabricate a stand-off, non-contact, microwave core temperature sensing system, but its accuracy and reliability will suffer from a much lower signal-to-noise ratio than a skin contact temperature sensing system, and a greater tissue-antenna impedance matching problem. The technology for a noninvasive core temperature measurement system is not mature enough to be incorporated in the engineering development (6.3) of a vital signs monitor. A simple skin contact microwave reflectometry system can be fabricated for about $15,000.00. Research and development in the areas of a very sensitive microwave system for noncontact core temperature measurement is still necessary.

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